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COMPUTERIZED BASELINE ESTIMATION AND AVERAGING OF EXERCISE ELEC--ETC(U)
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Report SAM-TR- 77-10

COMPUTERIZED BASELINE ESTIMATION AND AVERAGING OF EXERCISE ELECTROCARDIOGRAMS

June 1977

Final Report for Period March-October 1976

USAF SCHOOL OF AEROSPACE MEDICINE
Aerospace Medical Division (AFSC)
Brooks Air Force Base, Texas 78235

NOTICES

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This technical report has been reviewed and is approved for publication.

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REPORT DOCUMENTATION PAGE

READ INSTRUCTIONS BEFORE COMPLETING FORM

1. REPORT NUMBER
(14) SAM-TR-77-10

2. GOVT ACCESSION NO.

3. RECIPIENT'S CATALOG NUMBER

4. TITLE (and Subtitle)
(6) COMPUTERIZED BASELINE ESTIMATION AND AVERAGING OF EXERCISE ELECTROCARDIOGRAMS,

5. TYPE OF REPORT & PERIOD COVERED
(9) Final rept.
Mar[redacted]-Oct[redacted] 1976,

6. PERFORMING ORG. REPORT NUMBER

7. AUTHOR(s)
(10) Harold N. Keiser M.S.
Charles R. Meyer Capt, USAF

8. CONTRACT OR GRANT NUMBER(s)

9. PERFORMING ORGANIZATION NAME AND ADDRESS
USAF School of Aerospace Medicine (BRP, NGI)
Aerospace Medical Division (AFSC)
Brooks Air Force Base, Texas 78235

10. PROGRAM ELEMENT, PROJECT, TASK AREA & WORK UNIT NUMBERS
62202F
(16) 7755-18-06 (17) 18

11. CONTROLLING OFFICE NAME AND ADDRESS
USAF School of Aerospace Medicine (BRP, NGI)
Aerospace Medical Division (AFSC)
Brooks Air Force Base, Texas 78235

12. REPORT DATE
(11) June 1977

13. NUMBER OF PAGES
~~16~~ (12) 18p.

14. MONITORING AGENCY NAME & ADDRESS (if different from Controlling Office)

15. SECURITY CLASS. (of this report)
UNCLASSIFIED

15a. DECLASSIFICATION/DOWNGRADING SCHEDULE

16. DISTRIBUTION STATEMENT (of this Report)

Approved for public release; distribution unlimited.

17. DISTRIBUTION STATEMENT (of the abstract entered in Block 20, if different from Report)

18. SUPPLEMENTARY NOTES

19. KEY WORDS (Continue on reverse side if necessary and identify by block number)

Computer
Baseline
Electrocardiogram
Averaging

20. ABSTRACT (Continue on reverse side if necessary and identify by block number)

A cubic spline estimator of baseline variation and concepts of state-space can be combined to remove the baseline variation often encountered in stress electrocardiography. Since only one node, or knot, per cardiac cycle (the PR segment) is defined, baseline variation is removed with minimal effect on the low-frequency information within the cardiac cycle. The frequency response of the baseline estimator is a function of the heart rate.

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317000

COMPUTERIZED BASELINE ESTIMATION AND AVERAGING OF EXERCISE ELECTROCARDIOGRAMS

INTRODUCTION

When referred to the USAF School of Aerospace Medicine for evaluation, many aircrewmen have a treadmill exercise stress test performed. This test is used to determine the reaction of the subject's cardiovascular system to increasing levels of stress. During the entire testing period, the subject's vectorcardiogram is recorded on FM analog magnetic tape for use in future cardiovascular studies. The VCG is also recorded on a strip-chart recorder for the physician's use in evaluating the subject.

To prepare the data recorded on analog magnetic tape for future studies, each channel is digitized at 500 samples/sec by a Data General NOVA minicomputer. These data are stored as a series of digital values on digital magnetic tape, and the tape is processed on an IBM 360/65 to derive an averaged beat. Interval and rhythm measurements may then be made on the continuous data, while amplitudes and angle measurements are made on the averaged beat.

Almost all exercise vectorcardiographic data contain some high-frequency noise due to muscle activity and other exercise induced electrical activity. The data may also contain other types of extraneous electrical interference. To increase the signal-to-noise ratio, waves are averaged after a zero reference has been defined to remove low-frequency variations induced by respiration, perspiration, and other causes.

Typical solutions to the baseline-variation problem include (1) eliminating from future processing all beats whose zero-reference lines exceed a conservative threshold, (2) frequency spectrum filtering, and (3) first-order estimation with removal of the zero-reference values (4, 6). For an investigator taking measurements on individual beats or using a large volume of data, elimination of certain beats may be an efficient solution. Not all processing schemes can accept noncontinuous data, however, nor do all data sets have an abundance of beats. For these cases, filtering and zero-reference adjustment are requirements. Our experience in treadmill stress testing has indicated that a high-pass filter must have the lower 3-dB frequency above 0.5 Hz in order to remove often-encountered baseline frequencies. This filter, however, is in conflict with American Heart Association standards and, if applied to all VCGs, influences diagnostically sensitive low-frequency components such as ST segment (1). In an effort to reduce this effect, other investigators have used a straight line between the pre-P-wave and post-T-wave period of each cycle as a zero-reference line (2, 3). This straight-line estimator preserves low-frequency heart activity; however, it can follow only very low frequency variation. To preserve low-frequency heart information while removing higher frequency variation, we have increased the order of the estimator from one to three and selected one point (node, or knot) per cardiac cycle through which the estimate must pass.

Although the order of the estimator polynomial is greater than one, the implementation remains simple and efficient using state-space concepts (5). We use the mean of the values of a segment of the PR interval of a vectorcardiogram as a zero-reference point because it is not difficult to locate and is always present, even at high heart rates when there is no interval between a T wave and the following P wave. (The PR segment locator is described in appendix A.) At each PR interval there is a node through which the estimator must pass. We choose a third-order estimator, cubic spline, to connect these nodes.

THEORY AND IMPLEMENTATION

On the interval $0,T_1$ (see Fig. 1), let the baseline estimate, $y(t)$, be a cubic polynomial of the type:

$$y(t) = y'''(0)t^3/6 + y''(0)t^2/2 + y'(0)t + y(0) \tag{1}$$

(first four terms in a Maclaurin series, where derivatives of order four and higher are zero everywhere). At $t = 0$, we are given that

$$y(0) = y_0 \tag{2}$$

and

$$y'(0) = y_0' \tag{3}$$

(The initialization problem is addressed in appendix B.) For equation 1 to be totally determined, it is necessary to find $y''(0)$ and $y'''(0)$. By choosing a new y'' and y''' at each knot, only y and y' are continuous functions over the complete ECG record.

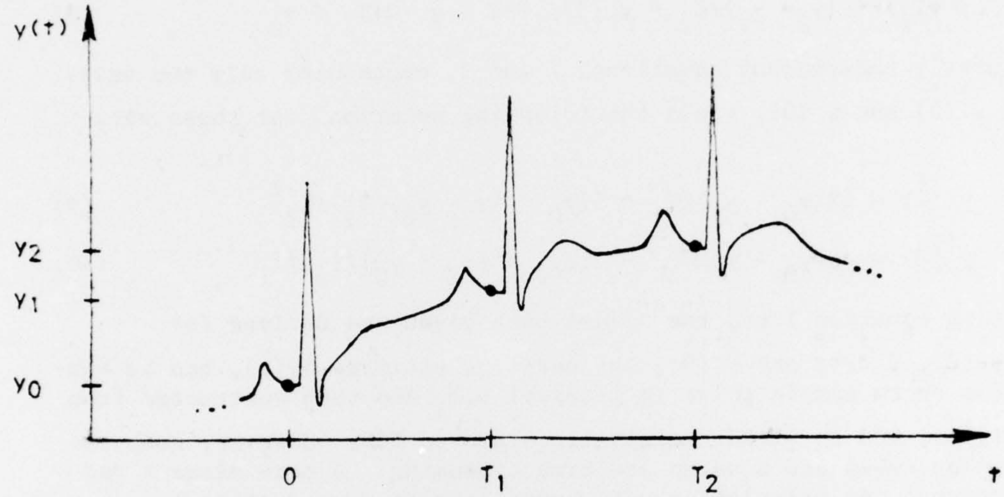

Figure 1. Typical ECG with PR knots identified.

An additional criterion that must be satisfied is that

$$y(T_1) = y_1 \tag{4}$$

e.g., the estimate must pass through the next node at sample T_1. Also, from stability considerations it seems prudent to choose

$$y'(T_1) = (y_2 - y_0)/T_2 \tag{5}$$

e.g., the estimate's slope at sample T_1 equals that of a straight line passing through the two adjacent PR segment knots. Classical splines of order three and higher, in which only the highest derivative is discontinuous, suffer stability problems during computation (5). But by defining both $y(t)$ and $y'(t)$ at each knot, we arrive at a stable solution. Note that over interval $0,T_1$, $y'(t)$ is determined by differentiating equation 1 with respect to time,

$$y'(t) = y'''(0)t_2 + y''(0)t + y'(0) \tag{6}$$

From equations 1, 2, 3, and 4, we get equation 7:

$$y_1 = y'''(0)T_1^{3}/6 + y''(0)T_1^{2}/2 + y_0' T_1 + y_0 \quad (7)$$

and from 3, 5, and 6, we get equation 8:

$$y'(T_1) = (y_2 - y_0)/T_2 = y'''(0)T_1^{2}/2 + y''(0)T_1 + y_0' \quad (8)$$

Two linearly independent equations, 7 and 8, containing only two variables, $y'''(0)$ and $y''(0)$, yield the following solutions for those variables:

$$y'''(0) = 12(y_0 - y_1)/T_1 + 6(y_0' + (y_2 - y_0)/T_2)/T_1^{2} \quad (9)$$

$$y''(0) = -6(y_0 - y_1)/T_1^{2} - 2(2y_0' + (y_2 - y_0)/T_2)/T_1 \quad (10)$$

Using equation 1 and the values both given and derived for $y(0)$, $y'(0)$, $y''(0)$, and $y'''(0)$, the baseline estimate, $y(t)$, can be computed for every sample point in interval $0,T_1$ and then subtracted from the original ECG to yield the baseline-removed ECG. However, computations using cubes and squares are time consuming. A more elegant and brief approach to calculating $y(t)$ between nodes uses a state-space (transition matrix) approach since the nonzero derivatives of $y(t)$ are limited to only the first three. Using equation 1 and differentiating with respect to time, we get the following system of equations, or state relationships:

$$y(t) = y'''(0)t^{3}/6 + y''(0)t^{2}/2 + y'(0)t + y(0)$$

$$y'(t) = y'''(0)t^{2}/2 + y''(0)t + y'(0)$$

$$y''(t) = y'''(0)t + y''(0)$$

$$y'''(t) = y'''(0)$$

and $y^{n}(t) = 0$ for $n \geq 4$.

In matrix notation

$$\begin{bmatrix} y(t) \\ y'(t) \\ y''(t) \\ y'''(t) \end{bmatrix} = \begin{bmatrix} 1 & t & t^{2}/2 & t^{3}/6 \\ 0 & 1 & t & t^{2}/2 \\ 0 & 0 & 1 & t \\ 0 & 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} y(0) \\ y'(0) \\ y''(0) \\ y'''(0) \end{bmatrix}$$

At this juncture, we still have not eliminated cubes or squares; however, by choosing t equal to one sample interval, we obtain:

$$\begin{bmatrix} y(1) \\ y'(1) \\ y''(1) \\ y'''(1) \end{bmatrix} = \begin{bmatrix} 1 & 1 & 1/2 & 1/6 \\ 0 & 1 & 1 & 1/2 \\ 0 & 0 & 1 & 1 \\ 0 & 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} y(0) \\ y'(0) \\ y''(0) \\ y'''(0) \end{bmatrix}$$

then by iterating, we get:

$$\begin{bmatrix} y(N+1) \\ y'(N+1) \\ y''(N+1) \\ y'''(N+1) \end{bmatrix} = \begin{bmatrix} 1 & 1 & 1/2 & 1/6 \\ 0 & 1 & 1 & 1/2 \\ 0 & 0 & 1 & 1 \\ 0 & 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} y(N) \\ y'(N) \\ y''(N) \\ y'''(N) \end{bmatrix}$$

where N and N+1 are adjacent sample indices. (N is a nonnegative integer.) Starting with $N = 0$, this system of equations can be recursively applied until all values of $y(t)$ have been computed on interval $0, T_1$. When computed, each $y(t)$ is subtracted from the original ECG data, and the result is stored back into the same data array location. In this manner, the baseline estimate is removed from the original ECG. At each new PR knot, a new $y''(0)$ and $y'''(0)$ are computed using equations 9 and 10, in which the abscissa of the new knot is reinitialized to zero; e.g., $N = 0$. The new values for y'' and y''' are inserted into the state column matrix:

$$\begin{matrix} y(0) \\ y'(0) \\ y''(0) \\ y'''(0) \end{matrix}$$

and the recursive process is begun again and continued until another knot is encountered.

This simple procedure is used to produce the average wave by defining fixed-length windows around the PR zero-reference points in the other windows. Each sum is then divided by the number of windows to provide the average wave which has the PR segment as its zero-reference point. These average waves can be measured for specific parameters upon which to base diagnostic or research findings.

RESULTS

Figure 2 illustrates the effects of our baseline estimation and removal process on an actual ECG lead obtained from a patient during treadmill exercise at the USAF School of Aerospace Medicine. The patient's ECG has an average heart rate of 110 beats per minute and contains both ST-segment depression and flattening. This original ECG is shown as the top tracing in Figure 2. The locations chosen for the PR knots, as described in appendix A, are marked by vertical lines. Computer-generated baseline noise (filtered Markov process with frequency components less than 0.5 Hz) was added to the original to create a baseline-corrupted ECG (the widely ranging ECG in Fig. 2). The baseline process (not the Markov noise that was added to originally produce the baseline process) was estimated as described previously in the Theory and Implementation section, and the processed ECG was obtained by subtracting the estimate from the baseline-corrupted ECG. The difference (error) between the patient's original ECG and the computer baseline-removed ECG, drawn to the same scale factor, is shown at the bottom of Figure 2.

Figure 3 illustrates the difference between the averages of the patient's original ECG and the baseline-removed ECG. Twenty beats were added to obtain each of these averages. The difference, shown at the same scale factor, is a straight line for all practical purposes.

Figures 4, 5, and 6 illustrate the frequency adaptation of the baseline-removal process. For these figures, the added baseline noise has frequency components above 1.0 Hz. The error magnitudes decrease with the increasing heart rate.

Figures 7 and 8 illustrate more quantitatively the effect of heart rate in removing a baseline noise of a given frequency. Shown at the top of the figures is purely sinusoidal baseline overlaid by its splinic estimate. Shown at the bottom is the error between the baseline and its estimate. The ECG is not important here and has been left out; only the knots at which the baseline process is sampled are shown. The significance lies in the ratio of the baseline frequency to baseline sampling rate. As can be seen in Figure 7, when the ratio of baseline frequency to baseline sampling frequency is 1/4, the error between baseline and estimate is less than 12%; and from Figure 8 when the ratio is 1/8, the error is less than 3%. In practice, the "baseline sampling frequency" is controlled by the heart rate of the individual being measured.

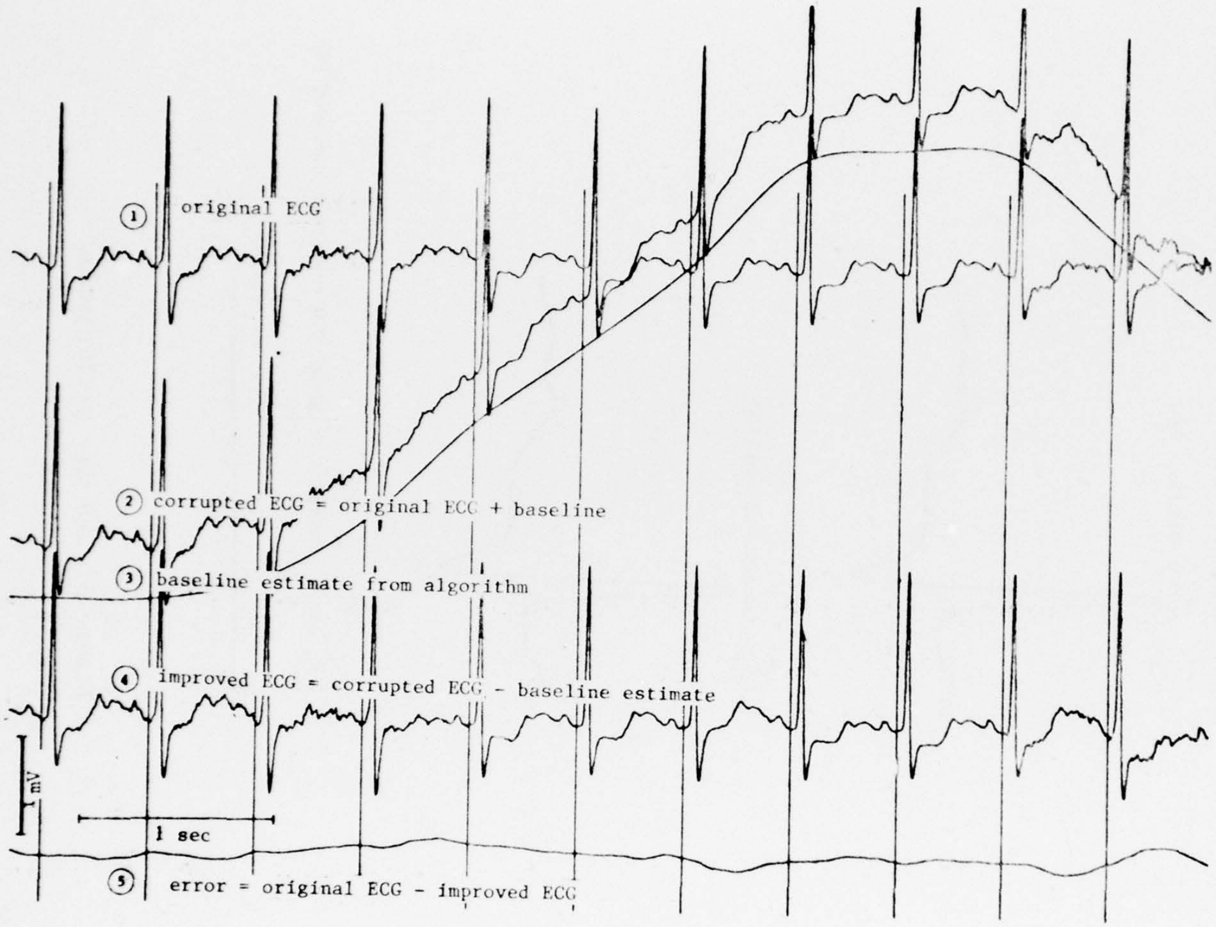

Figure 2. Performance of technique on a patient's ECG.

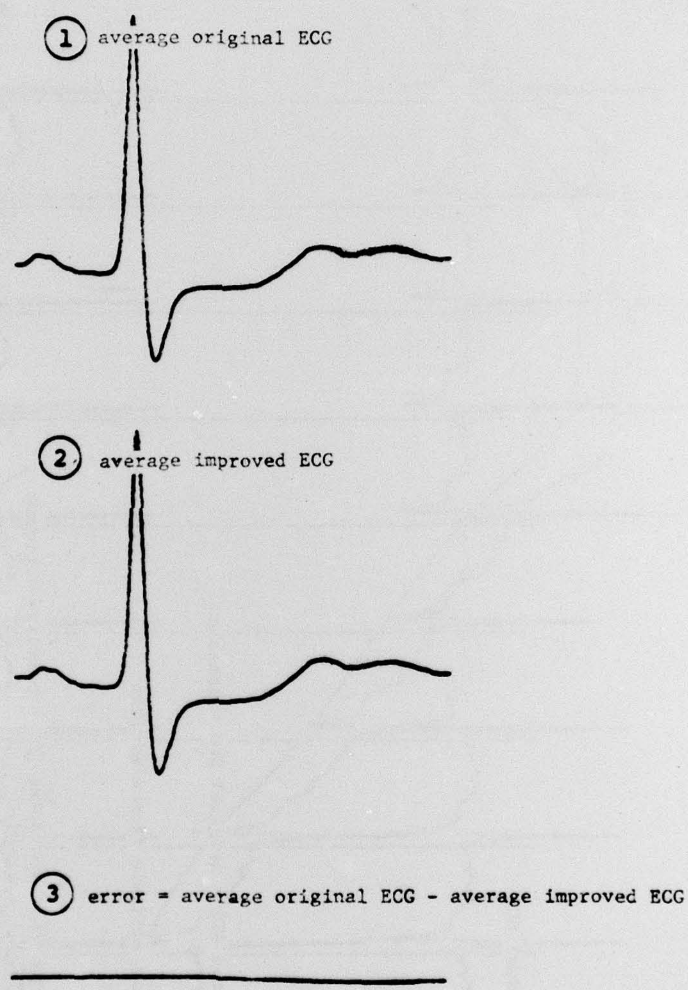

Figure 3. Average performance.

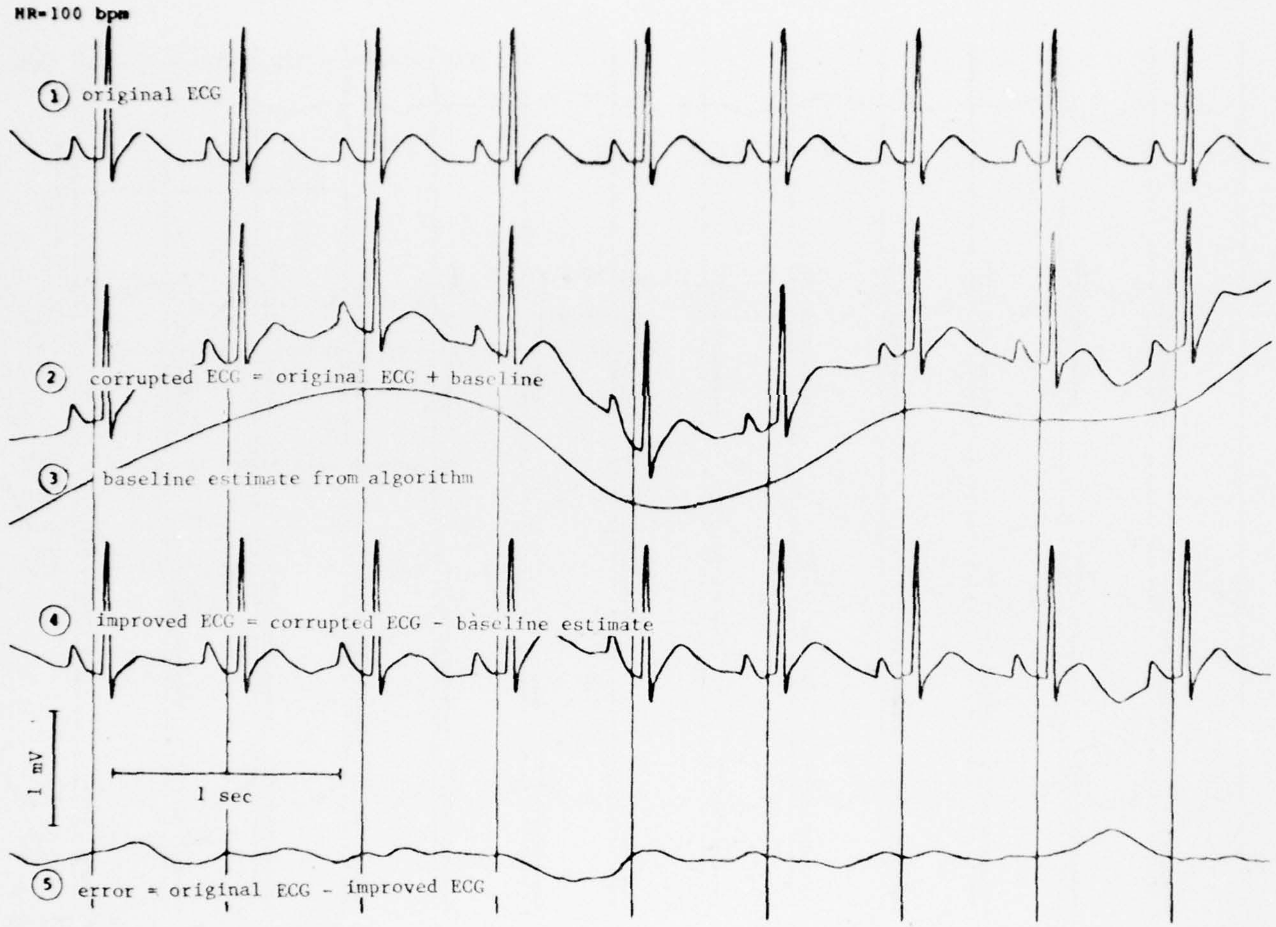

Figure 4. Performance of technique on synthetic ECG.

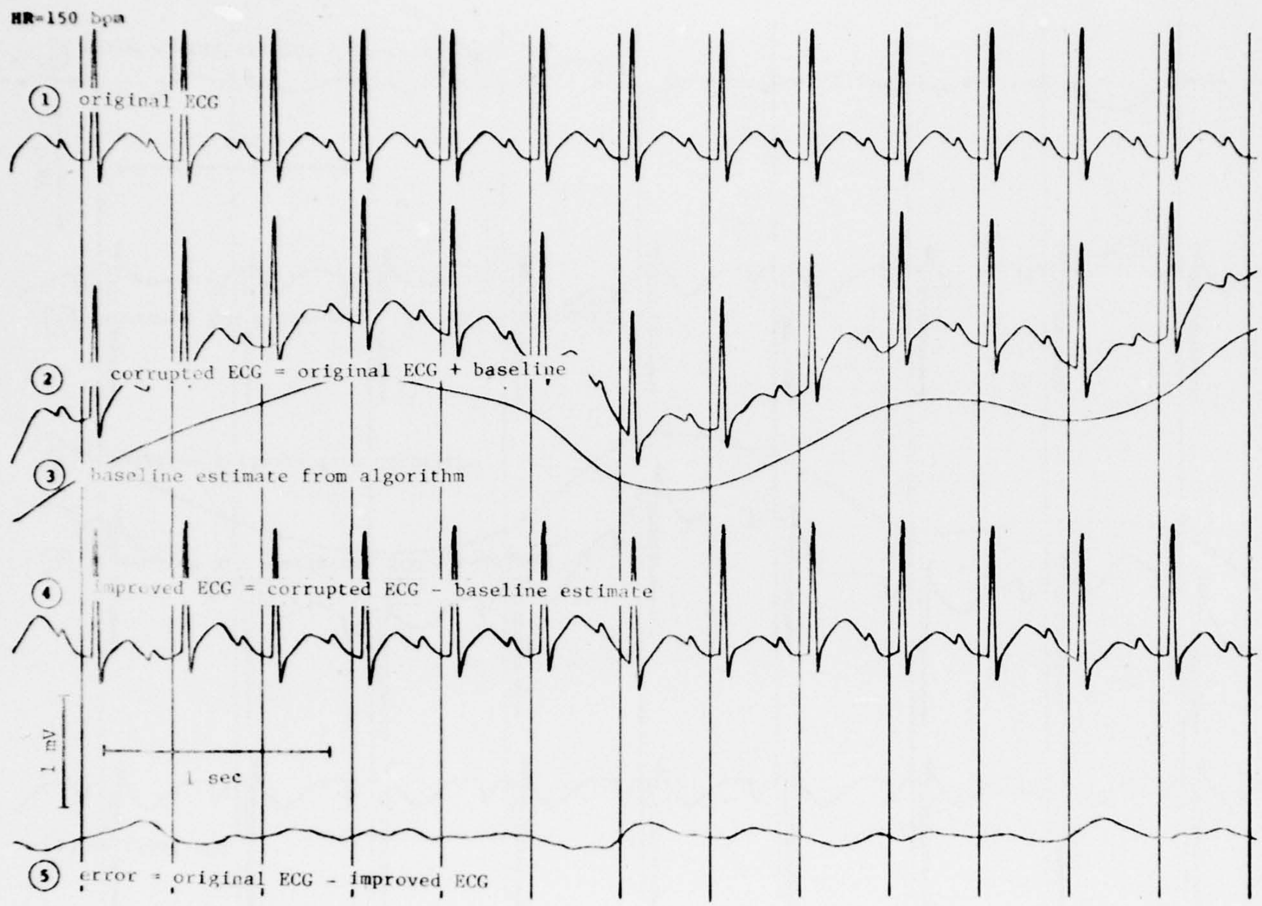

Figure 5. Performance of technique on synthetic ECG.

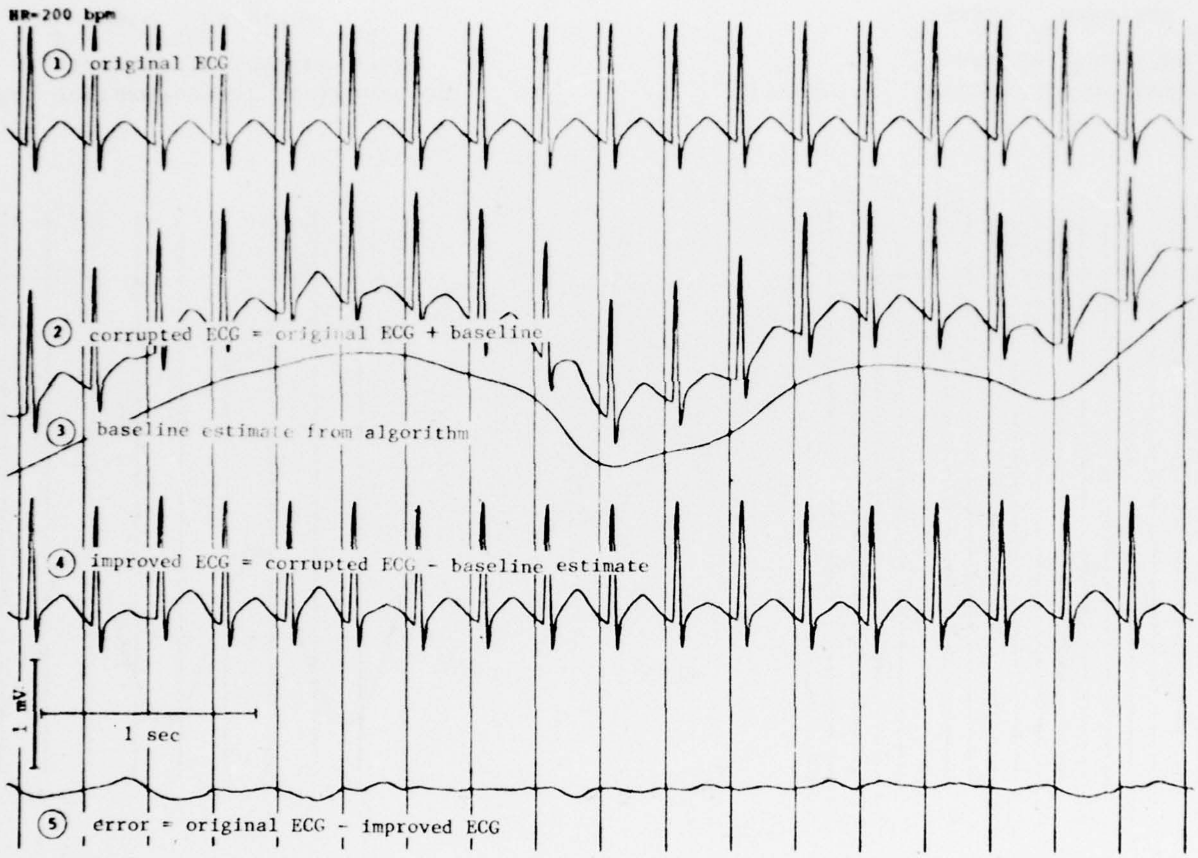

Figure 6. Performance of technique on synthetic ECG.

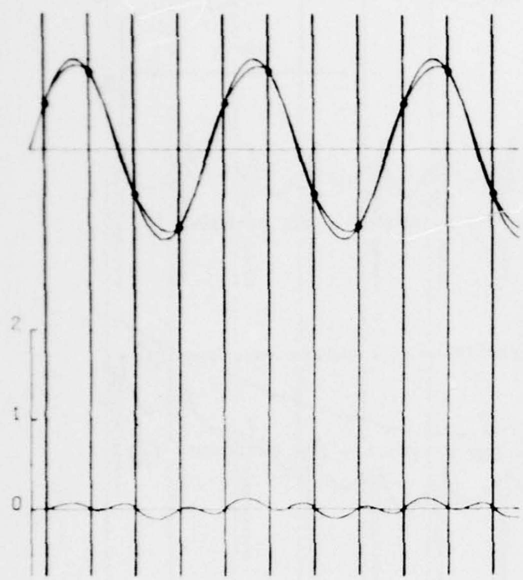

Figure 7. Performance of technique on sinusoidal baseline where $F_{sample}/F_{baseline} = 4$.

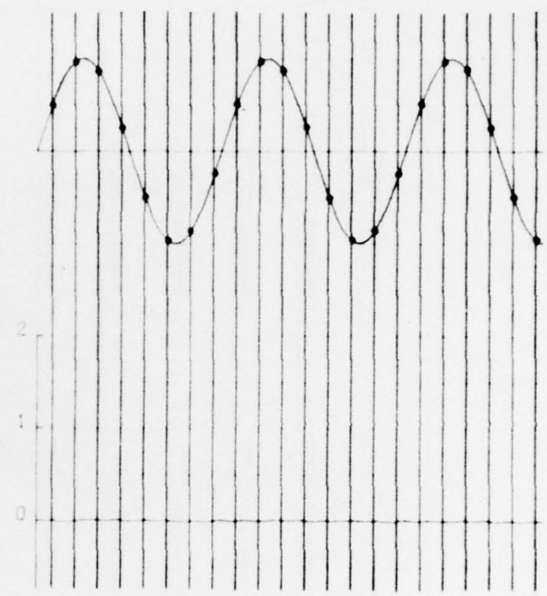

Figure 8. Performance of technique on sinusoidal baseline where $F_{sample}/F_{baseline} = 8$.

CONCLUSIONS

We have demonstrated a generalized and efficient method of using cubic splines to estimate the baseline process within an ECG; using only the PR-interval knots. The baseline is removed by simply subtracting the estimate from the raw data. Low-frequency heart activity is unaffected by this process since such activity is not admitted into the baseline estimate and hence cannot be subtracted from the raw data.

We have also empirically demonstrated that when the baseline sampling frequency is four times higher than a baseline frequency component, more than 88% of that baseline component is removed (i.e., at a resting heart rate of 60 beats per minute, baseline components at 0.25 Hz are effectively removed; and at an exercise heart rate of 150 beats per minute, baseline components up to 0.6 Hz are effectively removed without affecting ST segments).

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APPENDIX A

PR-INTERVAL KNOT IDENTIFICATION

The abscissa, or location in time, of a PR-interval knot is arbitrarily placed 66 msec before the R-wave's maximum downslope. The downslope of the ECG at any time I is computed using an average negative-slope estimate where

$$\text{downslope}(I) = \text{datum}(I-6) + \text{datum}(I-2) - \text{datum}(I+2) - \text{datum}(I+6)$$

The time interval between adjacent data points is 2 msec. The search for maximum downslope occurs when the computer downslope value exceeds 60% of the previous maximum (this 60% threshold may be initialized by calculating the maximum within the first 2 sec of data). During the search, the first downslope value that is less than its predecessor, defines its predecessor's value as the new maximum.

Once the abscissa of the PR-knot is located via the previous procedure, the ordinal value is chosen to be the average ordinal value of the 9 data points whose abscissae are nearest to and include the abscissa of the PR knot. Rationale for using 9 points to estimate the PR-interval knot, centers on eliminating the effects of 60-Hz noise based on the data-sampling frequency. An average over 9 points originally acquired at a sampling frequency of 500 Hz, spans 16 msec, or nearly one cycle of 60-Hz noise. From elementary digital-filtering theory, we know that averages consisting of symmetrically spaced points spreading exactly over one cycle of a sinusoidal signal are not biased by that signal component. Thus, a baseline estimate constructed from 9-point averaged PR knots where the original data sampling rate was 500 Hz, is relatively insensitive to 60-Hz noise.

APPENDIX B

PROCESS INITIALIZATION

The process may begin any number of ways. Two methods are shown below, both using the first three PR-interval knots.

1. This method simply defines a parabola through the first three knots to specify the initial state vector

$$\begin{matrix} y'''(0) \\ y''(0) \\ y'(0) \\ y(0) \end{matrix}$$

By algebraic manipulation, we determine that

$$y''(0) = 2\left[y_0(T_2 - T_1) - y_1T_2 + y_2T_1\right]/D$$

and $$y'(0) = -\left[y_0(T_2^2 - T_1^2) + y_1T_2^2 - y_2T_1^2\right]/D$$

where $$D = (T_1T_2^2 - T_1^2T_2)$$

and where $y'''(0) = 0$ for a parabola (second-order equation) and

$$y(0) = y_0$$

2. This method of beginning is to arbitrarily choose

$$y'(0) = (y_1 - y_0)/T_1$$

and

$$y(0) = y_0$$

Now by using text equations 9 and 10, $y'''(0)$ and $y''(0)$ may be determined.

The results presented in this paper used the parabolic initialization method.